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Ceramic-based layer structures for biomechanical applications

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Abstract

A survey of recent advances in the analysis of ceramic-based layer structures for biomechanical applications is presented. Data on model layer systems, facilitating development of explicit fracture mechanics relations for predicting critical loads to produce lifetime-threatening damage, form the basis of the work.

Keywords: Biomechanical replacements; Ceramics; Crack prevention; Dental crowns; Fracture mechanics; Layer structures; Radial cracks

1. Introduction

Ceramic materials are finding increasing usage in a wide range of technological structures. This is especially so in the area of biomechanical replacements—dental crowns, hip and knee prostheses, heart valves, bone implants, etc.—where wear resistance, biocompatibility, chemical durability, and even aesthetics, are critical issues. Fig. 1 depicts some examples. Generally, biomechanical replacements include more than one material type—ceramic, metal and polymer—commonly in some layer or other composite configuration. Whereas fatigue and wear in metal or polymer components can be limiting factors, ceramic components demand particular attention because of their brittleness. Yet despite an increasing incidence of catastrophic failures of ceramic-based prostheses in patients, the materials limitations of such devices are woefully understood by the medical community, where the *chief modus operandum* is the clinical trial. For the materials scientist, it is important to consider any prosthesis as a composite material system rather than a collection of individual monolithic components, with due attention to the mechanics of the human body. The states of loading can be complex, but the most common and most severe forms involve concentrated forces (P) from contacts of characteristic radius (r)—e.g. biting (teeth, $P \approx 100$ N, $r \approx 1$–10 mm) and body-weight support (hips, $P \approx 6000$ N, $r \approx 30$ mm). Prosthetic structures must be engineered to withstand such contacts, in exacting in vivo environments over millions of cycles.

In designing damage-resistant layer structures for any application it is important to distinguish between two (sometimes mutually exclusive) philosophies—crack containment and crack prevention. Virtually all the attention in the mechanics literature has focussed on crack containment. This philosophy is appropriate to large structures where the goal is to inhibit the penetration of existing cracks, either by enhancing crack deflection along weak interlayer interfaces to increase composite toughness [1,2], by incorporating residual compressive stresses in the ceramic layers to inhibit transverse crack growth [3,4], or by incorporating tough sublayers to arrest any penetrant cracks [3,5–10]. Crack prevention is more appropriate to smaller structures where the slightest damage may signal the end of safe function. This second philosophy applies to most biomechanical structures, and will consequently receive the bulk of our attention here. The problem is exacerbated in prolonged or cyclic loading, where small-scale damage can evolve steadily but inexorably over time into catastrophic failure [11,12].

In this article we survey recent studies on model ceramic-based laminate systems as a first step toward a fundamental understanding of the lifetime properties of biomechanical structures, using Hertzian contacts with spheres as a representative concentrated loading [**13]. Apart from its uniquely simple test configuration, Hertzian contact simulates the basic elements of occlusal [14–17,*18,**19] and hip [20] function. We present explicit

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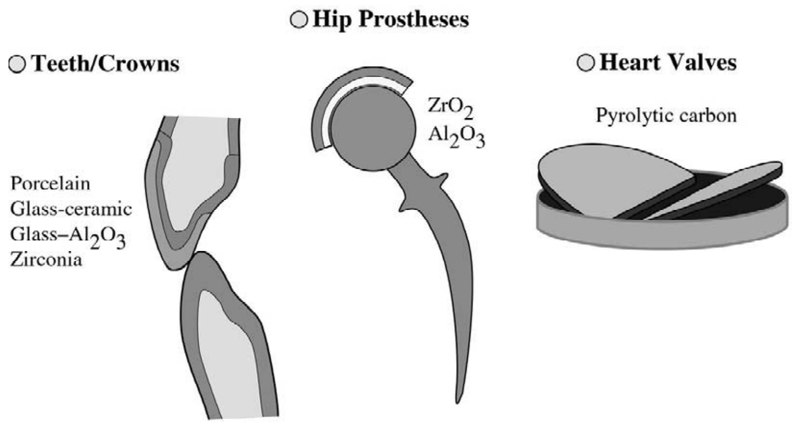

Fig. 1. Schematic showing selected biomedical replacements, indicating ceramic components: (a) dental crown, (b) hip prosthesis, (c) heart valve.

relations for the critical loads to produce different damage modes, in terms of basic material properties (modulus, strength, toughness and hardness) and geometrical quantities (layer thickness, contact radius). Results of experiments on simple bilayer, trilayer and multilayer systems are presented as validation of these relations. Guidelines for designing optimal material combinations are considered.

2. Damage modes

In this section we demonstrate damage modes in three clinically relevant bilayer structures, from indentation with WC spheres of radius $r = 2–4$ mm (Fig. 2): (a) micaceous glass-ceramic bonded with dental cement to a filled-polymer composite substrate; (b) porcelain fused to Pd-alloy metal; (c) porcelain fused to glass-infiltrated alumina. The figure shows top and side views obtained from 'bonded-interface' section specimens [21]. In each case the veneer coating layer is a brittle ceramic with properties similar to those of dental enamel; the substrate is representative of either compliant tooth dentin or hard crown core support material. Top-surface cone cracks in the coatings are observed in all three examples. Note that the cone diameters in Fig. 2a are wider than those in Fig. 2b and c. This is attributable to enhanced flexure of the ceramic plate on the softer polymeric substrate, shifting the maximum in surface tensile stress from the edge of the contact to the outer shoulders of the deflecting plate [**22]. Upward extending radial cracks are evident in Fig. 2a, indicating concurrent development of substantial tensile stresses at the lower plate surface [**22]. Radial cracks are also evident in Fig. 2b. In this instance the metal support, while stiffer than the porcelain, is also softer, facilitating substrate yield below the contact zone. Such yield allows the upper coating to deflect locally, with ensuing radial crack initiation [7,8]. No radial cracking is observed in Fig. 2c—the combined high stiffness and hardness of the alumina provides a more rigid support and precludes coating flexure. At first sight, these observations would appear to favor stiff and hard ceramics like alumina for substrate materials. However, such conclusions do not extend unequivocally to the design of crown-like trilayers, as we shall see later.

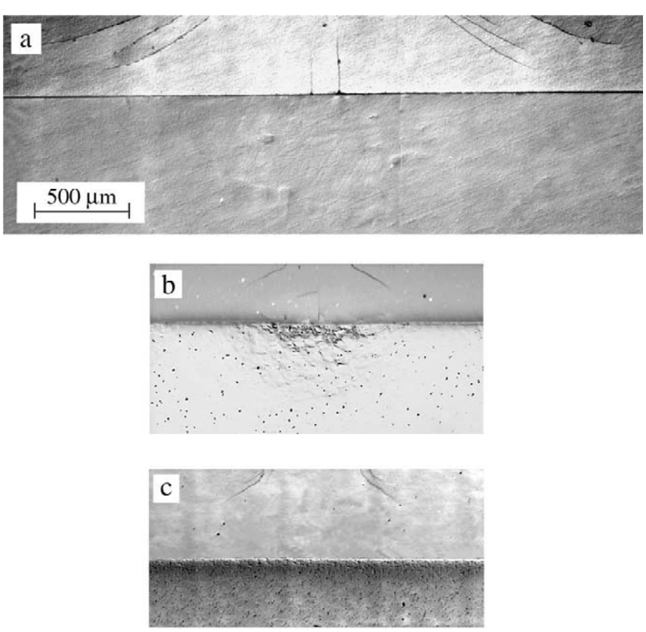

Fig. 2. Damage modes in selected sectioned ('bonded-interface') bilayers, from contact loading with WC spheres: (a) micaceous glass-ceramic bonded with dental cement to a filled-polymer composite substrate, $P = 450$ N [*35]; (b) porcelain fused to Pd-alloy metal, $P = 500$ N [*36]; (c) porcelain fused to glass-infiltrated alumina, $P = 500$ N [*35].

The competition between top- and lower-surface damage modes apparent in Fig. 2 is a general feature of ceramic overlayers on soft substrates. In highly brittle ceramics the spherical contact develops conventional cone cracks at the top surface [**13]. In less brittle, tougher ceramics, a second near-contact mode may operate—'quasiplasticity'—consisting of a 'yield' zone of distributed shear-microcracks [23,24]. Top-surface modes are favored in thicker, monolith-like coatings. At the bottom surface, radial cracks initiate. Radial cracks are believed to be the most common source of failure in dental crowns [*18]. They form more easily in thinner coatings, in flexure-like stress fields. Deformation of the substrate, either elastic or plastic, can be an important factor in facilitating the radial cracking mode.

3. Model layer structures

While section views of the kind shown in Fig. 2 are valuable for qualitative examination of contact-induced damage, they are severely limited in the quantitative information they can provide. Which cracks form first, and how does each evolve with increasing load to failure? Fracture mechanics analysis requires a more direct experimental approach.

A useful new route is the testing of model layer structures fabricated from transparent components—glass, sapphire, polycarbonate—for in situ observation of crack initiation and propagation [**22,*25,**26]. The layers are simply glued together with epoxy adhesive (also transparent). Viewing is performed either from below the polycarbonate substrate or through the glass or sapphire side walls. The glass and sapphire surfaces can be preabraded to control the strength properties in order to match dental polycrystalline counterparts (porcelain, alumina) and to selectively predetermine the site of cracking (i.e. top or bottom surface).

Fig. 3 shows examples of radial cracks in glass/polycarbonate bilayers and multilayers, viewed through the side walls. In both examples the radial cracks are in their well developed stages. Initiation has occurred at the glass undersurfaces, and the cracks have spread laterally and upward. In the four-layer specimen in (b), cracking occurred first in the top layer, then sequentially in the second, third and fourth layers, covering a range of almost a factor of six in the load ratio P_4/P_1. The radial cracks remain contained in their respective glass layers up to $P = P_4$, attesting to the damage tolerance of the structure.

Although the requirement of at least one transparent component for in situ viewing might appear to be restrictive, a wide range of model material systems can nevertheless be studied in this way: cracking in transparent ceramics on metal bases can be viewed from the side (as in Fig. 3) [*27]; radial cracking in the undersurfaces of opaque ceramic coatings can be observed through a transparent polycarbonate base from below [**28]; crown-like glass/sapphire trilayers can be viewed either from the side or from below [**26]. Provided the elastic mismatch between layers is sufficiently large, the newly initiated cracks tend to remain well contained within the brittle layers. Subsurface views reveal how the radial cracks grow stably outward from initiation sites close to the contact axis and multiply into expanding star-shaped configurations with increasing load [**22]. Most importantly, in situ viewing enables ready measurement of critical loads for crack initiation, especially radials, and thereby provides a sound basis for the establishment of a fracture mechanics analysis.

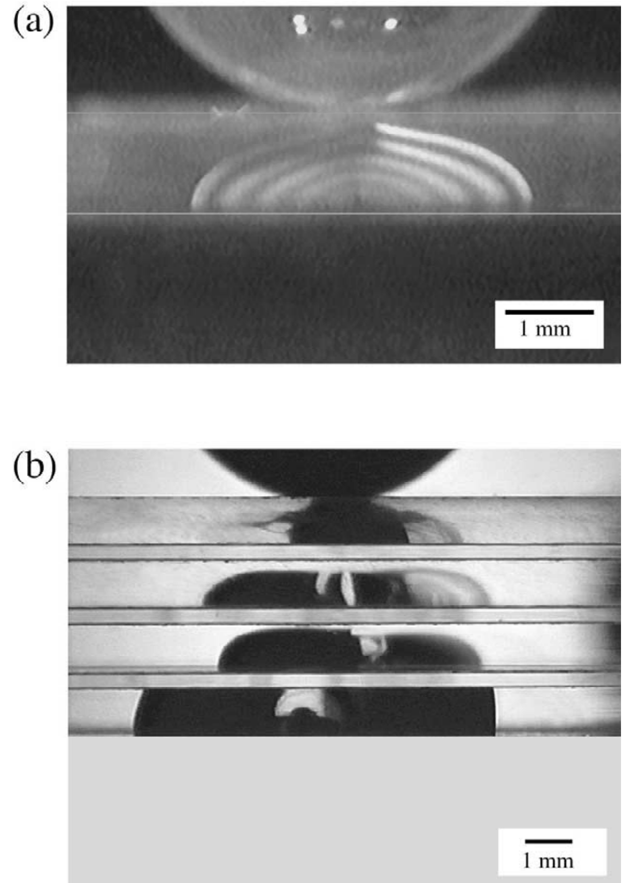

Fig. 3. Radial cracking in model layer systems fabricated by bonding glass slides to polycarbonate substrates and interlayers with epoxy adhesive, from contact loading with WC spheres. Glass bottom surfaces abraded to control strength and to selectively promote radial cracks at expense of surface damage. (a) Bilayers, contact load $P=130$ N. Fringes indicate interference at open crack surfaces. (b) Multilayers, $P=1370$ N. Critical loads for sequential formation of radial cracks: $P_1=230$ N, $P_2=850$ N, $P_3=1320$ N, $P_4=1370$ N. A cone crack has also become visible in the top glass surface.

4. Fracture mechanics

4.1. Bilayers

Closed-form relations expressing threshold conditions have recently been developed for cone crack (C), quasiplasticity (or yield, Y) and radial crack (R) damage modes in ceramic coatings of Young's modulus E and thickness d on compliant substrates of modulus E_s, subjected to contact with spheres of radius r at load P (Fig. 4a). For thick coatings, the critical loads for cone cracking and quasiplasticity at the top ceramic surface are given by [24,*29]

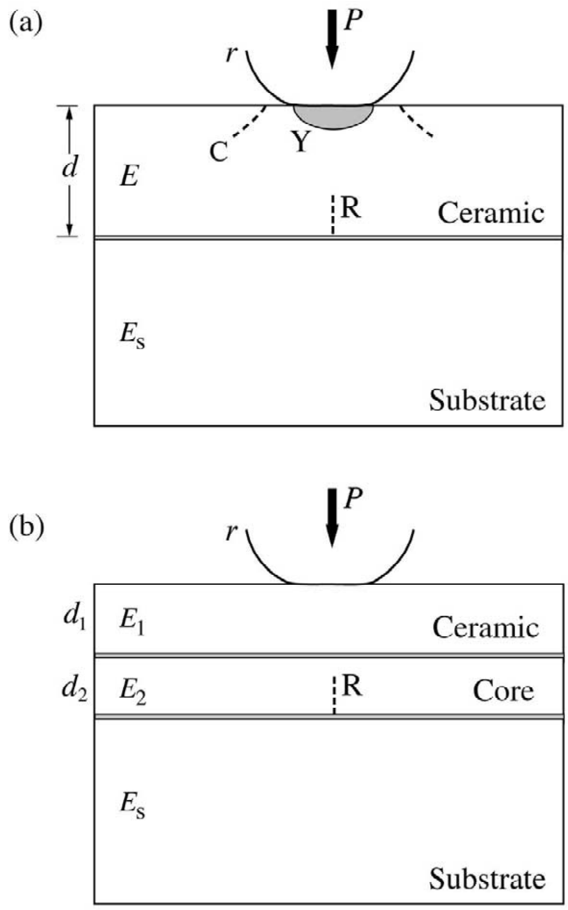

Fig. 4. Schematic of ceramic-based layer structures, (a) bilayer and (b) trilayer, indented with a sphere of radius r at load P. In trilayer, ceramic layer is replaced with ceramic veneer plus ceramic or metal core support, net layer thickness $d = d_1 + d_2$. Surface cone cracks (C) and quasiplastic yield zone (Y) initiate at top surface, radial cracks (R) at ceramic bottom surfaces.

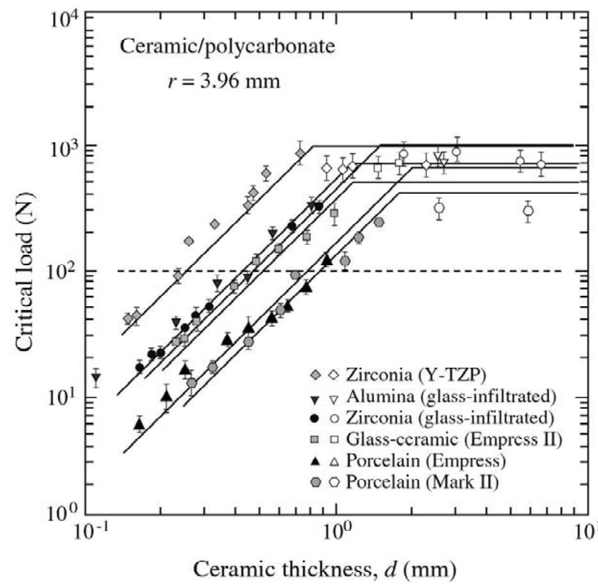

Fig. 5. Critical loads for first damage in ceramic/polycarbonate bilayers as a function of ceramic thickness d, for indentation with WC spheres of $r = 3.96$ mm, for a range of dental ceramics [**37]. Symbols are experimental data (standard deviation bounds). Solid lines are theoretical predictions for cone cracking and quasiplasticity (horizontal lines) and radial cracking (inclined lines). Dashed line indicates clinically relevant biting force.

$$P_C = A(T^2/E)r \tag{1a}$$

$$P_Y = DH(H/E)^2 r^2 \tag{1b}$$

with T toughness (K_{Ic}) and H hardness of the ceramic, and A and D dimensionless coefficients. For thin coatings, the critical load for radial cracking is given by [**22]

$$P_R = B\sigma_F d^2/\log(CE/E_s) \tag{2}$$

with σ_F the bulk flexural strength of the ceramic, and B and C as dimensionless constants [**28]. Note that P_C and P_Y in Eq. (1) are independent of d; similarly, P_R in Eq. (2) is independent of r. These remain reasonable approximations in the respective limits of d large or small.

Fig. 5 shows critical load data for selected dental-ceramic/polycarbonate bilayers. Points are experimental data: at large d (unfilled symbols), either cone cracking or quasiplasticity, whichever occurs first (quasiplasticity in all cases except Mark II porcelain); at small d (filled symbols), radial cracking. Solid lines are corresponding theoretical predictions from Eqs. (1) and (2). The dashed line at $P_m = 100$ N represents a nominal operational biting force in dental function. Safe design requires P_C, P_Y and (especially) $P_R > P_m$ in Eqs. (1) and (2). This may be realized by maintaining a conservatively large sphere radius and coating thickness ($r > 5$ mm and $d > 1.5$ mm, depending on the ceramic). From the materials standpoint, one seeks to maximize the ceramic parameters T, H and σ_F (not easily achieved simultaneously in any single ceramic). We note from Fig. 5 that Y-TZP zirconia is generally the most resistant to damage, porcelains the least resistant.

4.2. Trilayers

The most recent work has been aimed at extending the above analysis to trilayers, pertinent to all-ceramic dental crowns [**26]. Essentially, the monolithic ceramic coating is replaced by a bilayer consisting of an outer ('veneer') ceramic layer on a relatively stiff inner ceramic ('core') support layer (Fig. 4b). The top surface of the veneer layer is still subject to cone cracking (or quasiplasticity), as per Eq. (1). Both veneer and core are subject to radial cracking; but the core is especially susceptible because it sustains the bulk of the plate-like flexural stress, with maximum value at its undersurface. Thus it is the core and not the (usually weaker) veneer that demands closest attention in the context of design. The critical load for core radial cracking is expected to have a form analogous to Eq. (2), i.e.

$$P_R = B^* \sigma_2 d^2 / \log(CE^*/E_s) \quad (3)$$

with modified coefficient $B^* = B^*(d_1/d_2,\ E_1/E_2)$ and 'effective modulus' $E^* = E^*(d_1/d_2,\ E_1/E_2)$, and σ_2 the strength of the core material. Once the functions $B^*(d_1/d_2,\ E_1/E_2)$ and $E^*(d_1/d_2,\ E_1/E_2)$ have been determined, it should be possible to predict a priori the response of all-ceramic crown-like structures directly from bilayer data (e.g. Fig. 5). From these considerations, it may be expected that use of a stronger support material (e.g. Y-TZP) could lead to improved performance.

Fig. 6 is a plot of the critical load P_R to produce radial cracks in the core layers of glass/alumina/polycarbonate structures, as a function of d_2 (or d_1) at fixed $d = d_1 + d_2 = 1.5$ mm (nominal thickness of dental crowns). Relative to the limiting value for monolithic alumina coatings ($d_2 = 1.5$ mm, $d_1 = 0$), the critical load P_R diminishes as more of the core is replaced by glass veneer. In the context of dental crowns, that is the price of aesthetics. Note that the P_R data plateau out within $d_2 = 0.5$ to 1 mm, suggesting that the integrity of the structure is not too sensitive to the relative value d_2/d_1 in this intermediate region. However, P_R remains sensitive to the absolute value d_2 at any fixed d_1, via the d^2 dependence in Eq. (3)—recall the elimination of radial cracks in the bilayer limit $d_2 \rightarrow \infty$ (Fig. 2c).

Parallel studies are underway on ceramic/metal/polymer trilayers, in analogy to traditional porcelain-fused-to-metal dental crowns [*27]. The use of metal support layers would appear to eliminate the prospect of fracture in the core. However, as foreshadowed in Fig. 2b, the metal can yield, causing the ceramic overlayer to flex and generate radial cracks. For relatively thick cores ($d_2/d_1 > 1$), such that the location of first yield remains in the metal top surface, the critical load P_Y for yield is given to reasonable approximation by [*30]

$$P_Y = G H_m d_1^2 \quad (4)$$

with H_m the hardness of the metal and $G = G(E_1/E_2)$. Since core yield is an essential precursor to veneer radial cracking in these structures (i.e. $P_Y < P_R$), it is H_m that is the limiting material parameter. For thin cores, the metal flexes, and the location of first yield shifts to the metal bottom surface; the relationship between P_Y and material properties is then more complex.

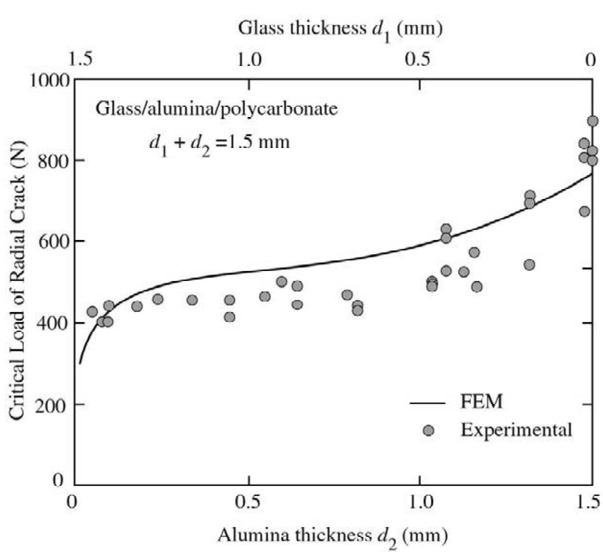

Fig. 6. Critical loads for radial cracking in the alumina core layer of glass/alumina/polycarbonate trilayers, as a function of alumina thickness d_2 (lower axis) or glass thickness d_1 (upper axis), for $d = d_1 + d_2 = 1.5$ mm. Data points are individual experimental results, solid curve is FEM calculation (courtesy Yan Deng and Pedro Miranda).

4.3. Multilayers

Finally, for multilayers of the kind represented in Fig. 3b, n like ceramic layers of fixed thickness d bonded with compliant interlayers of fixed thickness h, a semi-empirical extension of Eq. (2) for radial cracking in the uppermost layer (layer 1) relative to that in the lowermost layer (layer n) has been derived [*31]:

$$P_1/P_n = (1/n^2)[1 + \beta(d/h)^\gamma][\log(CE/E_s)/\log(CE/E_i)] \quad (5)$$

with β and γ dimensionless constants, subscripts s and i denoting interlayer and substrate. As indicated above, one of the advantages of using a multilayer instead of a monolayer coating is that any cracks tend to be constrained within individual layers. However, that advantage will be lost if P_1 becomes lower than P_n, from enhanced flexure on the soft interlayers [*25]. From Eq. (5), we see that this may be avoided by keeping n, h/d and E/E_i small.

5. Conclusions

In order to improve the lifetimes of biomechanical replacement structures, it is critical that we have a better understanding of material limitations. There is a compelling need for materials science input into a discipline traditionally governed by the mentality of the clinical trial. The present paper has sought to provide sound physical guidelines for predicting the onset of lifetime-threatening damage in representative biomechanical layer structures. Although we have focussed on dental crowns as a case study, the methodology is quite general. An emphasis has been placed on preventing cracks rather than containing them, by always operating in the elastic domain. Tests on model flat-layer specimens with transparent components for in situ viewing during contact loading provide an experimental basis for establishing explicit fracture mechanics relations that predict the onset of damage modes, particularly radial cracking, in terms of basic materials properties and key layer thickness variables. This approach enables quantitative rating of candidate materials for specific applications (e.g. Fig. 5).

There are many other factors that may contribute to the general performance of ceramic-based materials in layer structures. Fatigue processes may operate in repeat loading, from slow crack growth in the ceramic layers [32], quasiplastic damage accumulation in the ceramic layers [*33] (or plasticity in metal core layers), and viscoelastic processes in polymeric-based substrates. Flaw statistics can modify the critical load relations for radial cracking in Eqs. (2) and (3) by restricting the availability of large flaws at the ceramic undersurfaces [**34]. Residual stresses within individual layers may develop during fabrication and function [3], affecting the critical loads for damage. These factors are all amenable to incorporation into the fracture mechanics framework presented above. Complicating geometrical factors, such as occlusal convolution in dental crowns and departures from sphericity in hip prostheses, may be best handled by empirical FEM analyses or by testing in biosimulation machines. Biological interactions that limit the lifetimes of prostheses in vivo, for example the production of wear debris in total hip replacements, emphasize the need for any materials analysis to be considered in conjunction with clinical data.

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